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Biomechanical gait evaluation of the immediate effect of orthotic treatment for flexible flat foot

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Abstract

Flexible flat foot subjects attending the prosthetics and orthotics units come with prescriptions from orthopaedic surgeons for arch supports. Usually a pair of thermoformed plastic inserts are fabricated and fitted to treat the patients. However the effect of the orthotic treatment is not yet clear. A motion analysis system with two video cameras placed on the lateral and rear sides of the subject together with one force platform was used to investigate the immediate effects of the orthotic treatment. The force platform collected force data and the two cameras captured two-dimensional displacement data of the lower limb.

Eight subjects, all having an arch index (AI) larger than 3.0, participated in the study. For each subject, three successful steps on the force platform were videotaped for both the shod (with shoe only) and the orthotic (with shoe and orthosis) conditions. The kinetic variables were normalized to individual body weight and averaged for each subject. A Paired t-test was conducted to analyse sample means of matched pairs between the shod and the orthotic conditions.

The results showed changes in displacement data with relatively little change in the collected force data. The modified UCBL shoe insert evaluated significantly affected the orientation and movements of the subtalar joint, ankle joint and knee joint. These immediate effects reduced the degree and duration of abnormal pronation during the stance phase and thus had the potential for decreasing strain in the plantar ligaments and reducing abnormal tibial rotation which may be therapeutic for the foot.

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Introduction

Flexible flat foot is one of the most common lower limb conditions in children. Under weight bearing conditions the medial longitudinal arch of a flexible foot is depressed and the subtalar joint is pronated with the calcaneus assuming a valgus position. When weight bearing ceases, the arch remodels to a slightly more arciform shape compared to the loaded situation. There is a great deal of controversy regarding the management of this condition including whether it should be treated or not. Some authors have suggested that the flexible flat foot in young children will be self-correcting, requiring no treatment (Wenger *et al.*, 1989) except for those with congenital defects or neurological problems. In contrast, others suggest that those subjects who fall outside the normal range of parameters require some form of treatment (Rose *et al.*, 1985).

The subtalar joint of a normal foot pronates during the first 25% of the stance phase (Mann, 1982). This unlocks the midtarsal joints and allows the foot to adapt to uneven terrain. Following pronation, supination of the subtalar joint occurs, reaching its neutral position at mid-stance. The foot further supinates to become a rigid lever during the push-off phase. In the case of flexible flat foot, the subtalar joint remains pronated after foot flat. The midtarsal joint is not locked and the forefoot remains a mobile adapter instead of a rigid lever for propulsion. Resupination is delayed. Hypothetically during the push-off phase the pronation moment produced by the ground reaction force flattens the arch by rotating the subtalar joint. Thus the plantar ligaments have to resist the excessive and prolonged tension force caused by the abnormal range of pronation and the delay of resupination.

In addition to stretching the plantar ligaments, the excessive pronation of the subtalar joint produces prolonged internal rotation of the leg. This forces the patella laterally out of the patellar groove of the femur (Ramig *et al.*, 1980). In the patellofemoral articulation the patella normally slides smoothly over the groove in the anterior femur. With subtalar pronation, the patalla rides over the lateral aspect of the patellar groove and the patellar cartilage becomes irritated. Chondromalacia patellae, a painful condition of the knee, can be caused by this extrinsic biomechanical factor (Beckman, 1980).

Conservative management of patients with flexible flat foot is the recommended form of treatment by Lovell *et al.* (1986). Basmajian and Deluca (1985) concluded that muscle activity is not needed to support the arch of the fully loaded foot at rest but only when stress is applied, as at heel off, the aim of exercise is to strengthen the foot muscle only to prevent injuries that may be caused by ligamentous laxity. However in the presence of heelcord contracture, stretching exercises are preferred and orthoses are rarely indicated.

Various orthoses have been used in the management of flexible flat foot. These include a wide variety of corrective shoes, arch supports, and shoe inserts (Miller, 1990). These orthoses are mechanical devices designed to correct and maintain the foot near the optimum position so as to increase the efficiency of foot mechanics during walking or running and thus encourage normal development of the foot (Helfet, 1980). Based on static radiological data, Bordelon (1980), and Bleck and Berzins (1977) reported that significant correction of flexible flat foot deformity could be achieved by the use of orthoses. In contrast, Penneau *et al.*, (1982) and Wenger *et al.* (1989) suggested that the use of orthoses could not make permanent changes to the flexible flat foot. However none of the above reported on the functional outcome of the orthotic treatment.

This study aimed to quantify the immediate effect of orthotic treatment for flexible flat foot before establishing a longitudinal study to investigate long term outcomes. The specific objective of this study was to compare the kinetic and kinematic variables between the orthotic (with orthosis and shoe) and the shod (with shoe only) conditions with a view to evaluating the immediate biomechanical effects of the influences of the orthosis on the ankle-foot complex and the knee joint.

Method

Subjects

Eight subjects, 7 females and 1 male participated in this study. They were between the ages of 4 and 11 years (mean age for all subjects was 6.3 years) and were referred from the paediatric orthopaedic clinic of the Queen Elizabeth Hospital, Hong Kong. Each subject received a lower limb musculoskeletal examination. Measurement of range of motion of the ankle joint and subtalar joint was performed to ensure that they all had various degree of bilateral flexible flat feet that were not compensation of forefoot deformities or other confounding pathology.

Orthosis

Each subject was fitted with a pair of orthoses commonly used in Hong Kong based on the UCBL (University of California Biomechanics Laboratory) shoe insert design (Henderson and Campell, 1967; Kogler *et al.*, 1996). Since the partial weight bearing casting method suggested by Henderson and Campell was not considered ideal by the authors for obtaining a neutral cast of the foot with the subtalar joint in the neutral position, a non-weight-bearing prone casting method was employed (McPoil *et al.*, 1989). Following the application of plaster bandages to the foot to form the negative impression the orthotist used the thumb and index fingers to palpate the head of talus to ensure that the subtalar joint was in the neutral position. The midtarsal joint was fully locked by applying a dorsiflexion force through the thumb of the other hand, placed on the plantar surface of the fourth and fifth metatarsal heads, which dorsiflexed and abducted the forefoot. The negative impression was used to create a positive cast.

Modifications to the positive cast included

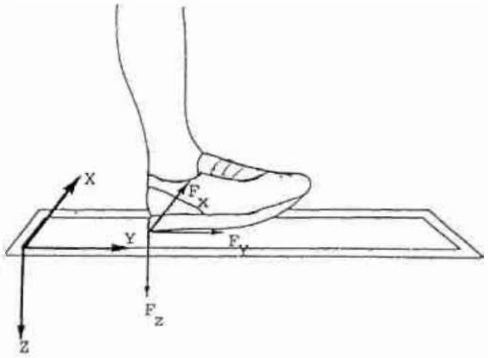

Fig. 1. Force components measured by the force platform.

those in the guidelines suggested by Henderson and Campbell (1967), and Colson and Berglund (1977): to include: (1) removal of plaster along the lateral aspect of the fifth metatarsal and the medial aspect of the head of the first metatarsal to match the measured width of the foot across the first and fifth metatarsal heads to control abduction and adduction of the forefoot; (2) removal of material from the lateral area of the heel to make it about 0.3 cm undersize in order to grip the calcaneus closely; (3) removal of material of about 0.8 cm in depth from the posterior aspect of the medial longitudinal arch so that it blended in with the medial support area above the calcaneal tuberosity in the area of the sustenaculum tali, for support of the calcaneus; and (4) addition of plaster, about 0.4 cm, at the navicular tuberosity to relieve pressure.

Following thermoforming with a 4 mm polypropylene sheet, the medial trimline was kept as high as comfort permitted while the lateral trimline was allowed to be considerably lower. The medial and lateral walls gripped the calcaneus to limit movement toward a valgus position. Under the plantar surface of the foot the distal trimline of the orthosis was located proximal to the heads of the metatarsals. The patients all reported that the orthoses were comfortable when initally fitted and at the follow-up appointment two weeks later.

Instrumentation

The biomechanical analysis of the effect of orthotic treatment for flexible flat foot was carried out through a two-dimensional video-based, computer-interfaced gait analysis system (Areblad *et al.*, 1990; Cornwall and McPoil, 1995) which consisted of a force plate and two video cameras. The force plate was located 5m from the start of the walkway, with the two cameras one at the start of the walkway, and the other 3 m to the right, in line with the force plate. The force plate (Advanced Mechanical Technology Inc., Newton, Massachusetts, USA) was used to measure force data: the vertical force (Fz), the anterior-posterior force (Fy) which was the horizontal force along the line of progression, the medial-lateral force (Fx) which was the horizontal force 90 degrees from the line of progression, and the axial torque measurement which was related to the reaction to the transverse rotation occurring in the lower limb during gait (Fig. 1). The Peak Motion Measurement System (Peak Performance, Inc., Englewood, Colorado, USA) was used for collection of displacement data. The force plate and the two cameras were synchronised. Both force and displacement data were sampled at 50 Hz.

Data collection

The arch index (Cavanagh and Rodgers, 1987) of each subject was calculated to quantify the degree of flatfoot (Table 1). A piece of paper was placed under an inked rubber differential pressure mat to collect the footprint of each subject during a step. When the subject stepped on it, the pressure pressed the ridges of the mat onto the paper and left an inked footprint on the paper. On the footprint (Fig. 2) a line is drawn from the centre of the heel (point X) to the tip of the second toe. This is called the

Table 1. Arch Index of the 8 subjects.

Subject	1	2	3	4	5	6	7	8
Sex	F	F	F	F	F	M	F	F
Age	11	11	11	9	4	5	4	5
Weight (N)	325	277.4	270	341	165.8	170.5	173	206
Arch Index	0.32	0.39	0.35	0.37	0.35	0.34	0.39	0.38

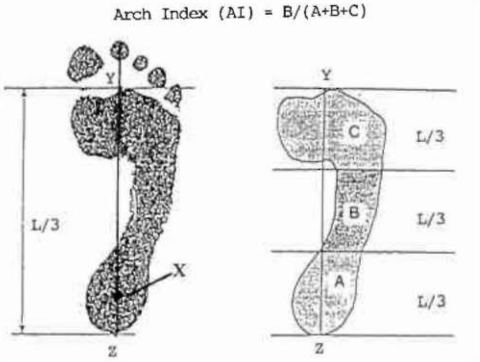

Fig. 2. Arch Index (AI) (after Cavanagh and Rodgers, 1987).

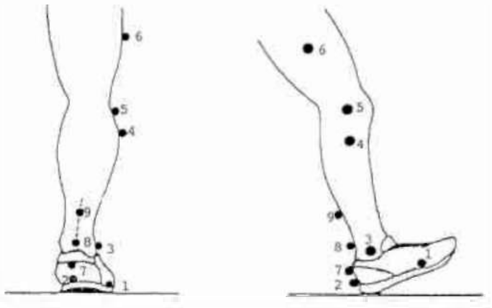

Fig. 3. Locations of the reflective markers. 1. MT5; 2. Cali; 3. LM; 4. FH; 5. Thi; 6. Ths; 7. Cals; 8. Legi; 9. Legs.

foot axis. A second line is drawn perpendicular to this axis such that it is tangential to the most anterior part of the outline of the main body of the footprint in front of the metatarsal heads. The intersection of these two lines is marked (point Y). The line XY is extended to the most posterior part of the heel at a point Z. The line YZ is divided into three equal parts dividing the foot, with the toes ignored, into rearfoot (A), midfoot (B) and forefoot (C). The Arch Index (AI) is the ratio of the area of the midfoot (B) to the total area (A+B+C) of the foot.

$$AI = B/(A+B+C)$$

To measure the motions of the knee and the ankle-foot complex in the sagittal plane and the coronal plane (Nigg, 1986; McCulloch *et al.*, 1993), spherical reflective markers of 1 cm diameter were attached, with double sided adhesive tape, to the right lower limb of each subject at the critical locations as follows:

(1) midsole of the shoe at the location of the head of the 5th metatarsal,
(2) the centre of the sole corresponding to the inferior bisector line of the calcaneus,
(3) the lateral malleolus,
(4) the head of fibula,
(5) the knee joint,
(6) a point proximal to the knee joint and two thirds of the distance on a line from the knee joint to the greater trochanter,
(7) the heel counter corresponding to the superior bisector line of the calcaneus,
(8) centre of Achilles tendon at the height of medial malleolus,
(9) 10 cm above marker of 8, at the centre of the leg as viewed from the rear, below the gastrocnemius heads (Fig. 3).

The reflective markers which appeared bright against the dark background were tracked by the motion analysis system. The components of the foot-floor direct force and the axial moment were measured by the force platform. The displacement data was synchronised with the force data. As the shoe is considered as part of the foot orthotic system and will affect the gait, standard laced leather shoes with firm heel counter and flexible sole were provided during the data collection sessions. Each subject was asked to walk at his/her normal speed with and without orthoses inside his/her shoes on both feet. Data collection trials were run until three successful trials for each condition, shod (without orthosis) and orthotic were obtained.

Data analysis

The raw displacement data were filtered with a Butterworth filter. The cut off frequency was 10 Hz. The conditioned data were then further reduced to obtain the scaled linear

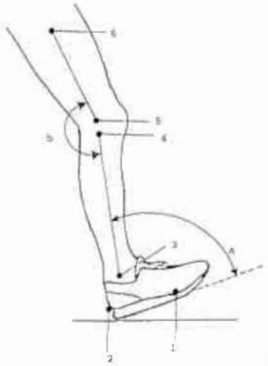

Fig. 4. Segmental angles of the lateral image. Angle a, ankle angle; Angle b, knee angle.

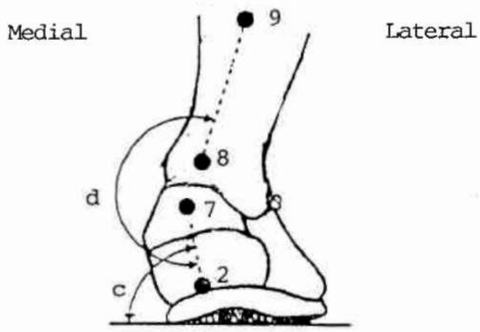

Fig. 5. Segmental angles of the rear image. Angle c, rearfoot angle; Angle d, eversion angle.

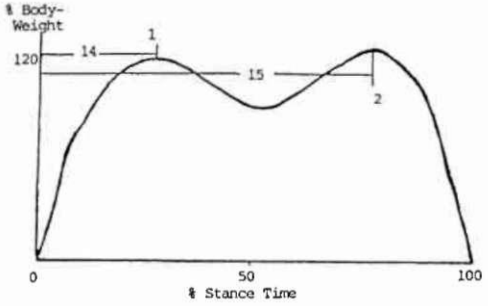

Fig. 6. Selected variables of vertical force component. 1. the first peak vertical force; 2. the second peak vertical force; 14. % stance at first peak vertical force; 15. % stance at second peak vertical force.

displacements of various markers and the angular displacements between line segments. The knee angle was defined as the posterior angle between the thigh and the lower leg in the sagittal plane, and the ankle angle was the anterior angle between the leg and the sole (Fig. 4). In the coronal plane, the rearfoot angle was the medial angle between the bisector line of the calcaneus and the horizontal, and the eversion angle was the medial angle between the bisector lines of the lower leg and the calcaneus (Fig. 5). To take into account the influence of the body weight on the recorded force data, the data was normalised by dividing by the individual subject's body weight. The kinetic and kinematic variables of the three successful trials were separately averaged for each subject, and group means were also calculated. Paired t-test was conducted to analyse sample means of matched pairs between the orthotic and non-orthotic conditions. Each subject served as his/her own control in determining if there was a treatment effect. Statistical significance was claimed for $p<0.05$.

To take into account the influence of the body weight on the recorded force data, the data was normalized by dividing by the individual subject's body weight. The variables (Hamill *et al.,* 1989) that were defined in the force-time data for analysis are listed in Table 2.

The variables (Fig. 9) that were defined in the displacement data in the frontal plane are listed in Table 3.

The variables (Fig. 10) that were defined in the displacement data in the sagittal plane are listed in Table 4.

Table 2. Variables that were defined in the force-time data.

Vertical force	(1)	First peak (Fig. 6)	(2)	Second peak (Fig. 6)
	(3)	Average		
Anterior-posterior force	(4)	Peak anterior (Fig. 7)	(5)	Peak posterior (Fig. 7)
	(6)	Average		
Medial-lateral force	(7)	First peak medial (Fig. 8)	(8)	Second peak medial (Fig. 8)
	(9)	First peak lateral (Fig. 8)	(10)	Second peak lateral (Fig. 8)
	(11)	Average		
Vertical torque	(12)	Peak external (Fig. 9)	(13)	Average
% Stance time at	(14)	First peak vertical force (Fig. 6)	(15)	Second peak vertical force (Fig. 6)
	(16)	Peak anterior Force (Fig. 7)	(17)	Peak posterior force (Fig. 7)
	(18)	Zero A-P force (Fig. 7)	(19)	First peak medial (Fig. 8)
	(20)	Second peak medial (Fig. 8)	(21)	First peak lateral (Fig. 8)
	(22)	Second peak lateral (Fig. 8)	(23)	Peak external torque

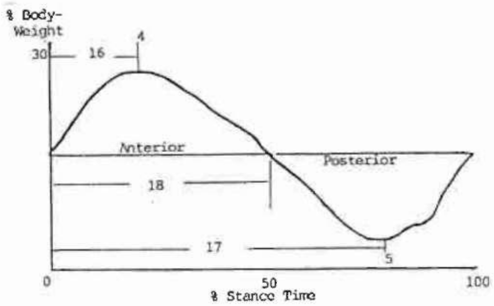

Fig. 7. Selected variables of anterior-posterior force component. 4. the peak anterior force; 5. the peak posterior force; 16. % stance time at peak anterior force; 17. % stance at peak posterior force; 18. % stance at zero anterior-posterior force.

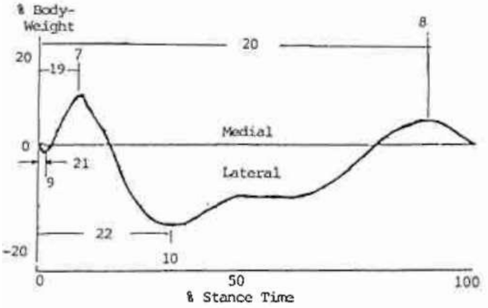

Fig. 8. Selected variables of the medial-lateral force component. 7. the first peak medial force; 8. the second peak medial force; 9. the first peak lateral force; 10. the second peak lateral force; 19. % stance at first peak medial force; 20. % stance at second peak medial force; 21. % stance at first peak lateral force; 22. % stance at second peak lateral force.

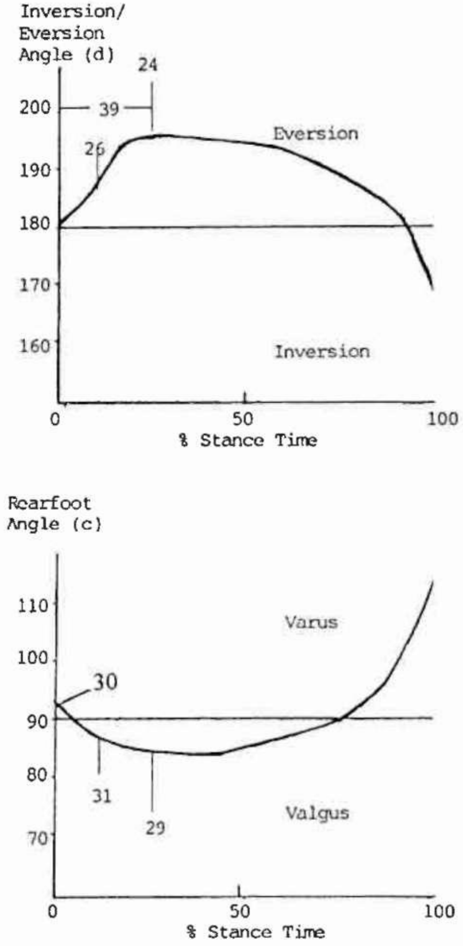

Fig. 9. Selected variables of the kinematic data of the posterior view. 24. the maximum eversion angle; 26. the eversion angle at 10% stance time; 39. % stance at maximum eversion angle. 29. the rearfoot angle at maximum eversion; 30. the rearfoot angle at heel strike; 31. the rearfoot angle at 10% stance time.

Table 3. Variables that were defined in the displacement data in the frontal plane.

Eversion Angle	(24) Maximum	(25) At heel strike
	(26) At 10% stance time	(27) Initial = (25) - (26)
	(28) Total = (24) + (25)	
Rearfoot angle	(29) At maximum eversion	(30) At heel strike
	(31) At 10% stance time	(32) Initial = (31) - (30)
	(33) Total change = (30) - (29)	
% Stance time at	(34) Maximum eversion	

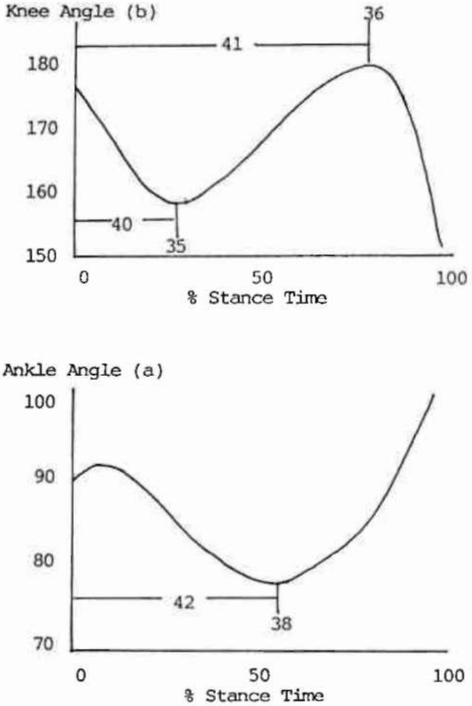

Fig. 10. Selected variables of the displacement data in the sagittal plane. 35. the maximum knee flexion angle; 36. the maximum knee extension angle; 40. % stance at maximum knee flexion angle; 41. % stance at maximum knee extension angle; 38 the ankle angle at maximum dorsiflexion; 42. % stance at maximum ankle dorsiflexion.

Results

Both the vertical force patterns of the shod and orthotic conditions were of the typical two peaks curves. There were no significant differences in the selected variables of the vertical force component. The anteroposterior force component also showed anterior and posterior forces of high degree of symmetry with no significant difference in the selected variables between the shod and the orthotic conditions. Significant differences ($p<0.05$) of the mean values between the orthotic condition and the shod condition were noted and are shown in Table 5.

Discussion

The results showed that selected variables of the vertical force components were not influenced by orthotic use. One of the functions of the pronation of the subtalar joint is to reduce the vertical impact force during the shock absorption phase which lasts from heel strike to foot flat (Berger *et al.*, 1981). The magnitude of the first peak of vertical force, the vertical impact peak force in the two conditions indicated that the shock absorption function was not affected by the use of the orthosis. The orthotic intervention also did not influence the timing of the vertical force components.

In the orthotic condition there was observed to be a 1.8% decrease ($p<0.05$) in the magnitude of the second peak lateral force and a 1.5% increase ($p<0.05$) in the magnitude of the average medial-lateral force over the stance phase i.e. to a less negative value. The delay of 2.5% in stance time ($p<0.05$) at the occurrence of peak external torque indicated the influence on the abduction movement of the foot. The significant changes in the two mediolateral variables might be because orthotic control of the foot reduces internal tibial rotation and resulted in a less abducted forefoot that exerted a lower lateral force. The delay in the occurrence of the external torque peak might also be due to the influence of the orthosis on the abduction movement of the foot. In the orthotic condition, with the exception of the degree of eversion at heel strike, there was reduction in all other eversion variables. The orthotic intervention reduced the maximum eversion by 37.1% and the total eversion by 39.5%. The difference of the initial eversion at heel strike between the shod and orthotic conditions was not significant. This non-significant difference did not show the foot orthosis had an effect on the subtalar joint

Table 4. Variables that were defined in the displacement data in the sagittal plane.

Knee angle	(35)	Maximum knee flexion	(36)	Maximum knee extension
	(37)	At maximum eversion		
Ankle angle	(38)	Maximum dorsiflexion	(39)	At maximum eversion
% Stance time at	(40)	Maximum knee flexion	(41)	Maximum knee extension
	(42)	Maximum dorsiflexion	(43)	Heel off

Table 5. Mean percentage difference of variables between the shod condition and the orthotic condition.

Variables	Description	Mean value	Standard deviation
2	second peak lateral force	-1.8%	2.3%
11	average medial-lateral force over the stance phase	1.5%	1%
23	stance time delay for the occurrence of the external torque peak	2.5%	2.7%
24	maximum eversion	-37.1%	9.1%
28	total eversion	-39.5%	10.8%
34	stance time delay for the occurrence of maximum eversion	16.2%	7.1%
29	rearfoot angle at maximum eversion	3.5%	1.1%
33	total rearfoot angular change	-38%	11.8%
37	knee angle at maximum eversion	2.4%	2.3%
38	ankle angle at maximum dorsiflexion	-2.8%	1.6%
42	atance time delay for the occurrence of the maximum dorsiflexion	4.8%	7%
39	ankle angle at maximum eversion	13.4%	6.8%
43	stance time delay for the occurrence of heel off	8.3%	3.8%

movement during the shock absorption phase. This would explain why the magnitude of the first peak vertical force was not significantly different in the shod and orthotic conditions. The percentage stance time at which the maximum eversion occured in the orthotic condition was reduced by 16.2%. This was much closer to the average value of 28% stance time than the 46.2% stance time in the shod condition. This suggests that, with the orthosis, the subtalar joint of the subject resupinated relatively earlier than that in the shod condition. Should this be the case abnormal stress to the ankle foot complex may be reduced. This presumably leads to a smaller magnitude of and shorter period of excessive tension on the plantar ligaments and also less internal rotation of the tibia, both in terms of duration and range of movement.

The rearfoot angle at heel strike did not change when the orthosis was introduced. As a factor of the eversion angle, the rearfoot angle at maximum eversion increased by 3.5% ($p<0.05$); and the total rearfoot angular change reduced by 38% ($p<0.05$). At maximum eversion, the knee angle increased by 2.4% ($p<0.05$). This agreed with the reduction in maximum eversion angle as supination of the subtalar joint moves the proximal aspect of the tibia backward, thus extending the knee (Root *et al.,* 1977). The reduction of maximum dorsiflexion angle at the ankle by 2.8% ($p<0.05$) would indicate an increase in dorsiflexion in the orthotic condition. The % stance time at maximum dorsiflexion was delayed by 4.8% ($p<0.05$). These indicated that the ankle motion occurred from a relatively supinated position. The ankle angle at maximum eversion increased by 13.4% ($p<0.05$) indicating a less dorsiflexed ankle at that time. The occurrence of heel-off was also 8.3% stance time delayed for the orthotic condition. This allowed a longer period for resupination following maximum pronation.

Conclusions and further research

Results of this study in the control of the motion of the subtalar, ankle and knee joints were comparable with the reports of previous studies on functional foot orthoses (McCulloch *et al.,* 1993; Novick and Kelley, 1990; Rodgers and LeVeau, 1982; Smith *et al.,* 1986) used to modify pronation during running and walking. The modified UCBL insert held the foot in a position that relieved tension on the soft structures, thus forming the arch without excessive pressure on it. The calcaneus was gripped by the medial and lateral walls of the orthosis and supported by the blended surface of the orthosis under the sustentaculum tali to limit movement toward a valgus position. The lateral wall of the orthosis also controlled the abduction of the forefoot at the mid-tarsal joint. The flexible flat foot was held and allowed to function and grow in appropriate alignment with less stressful force on the soft tissue. If the growing foot develops and functions in the shape in which it is held, in the long run, the ligament laxity would be reduced and the dynamic deformity would be corrected.

From the results of this experiment several gait parameters have been quantified that could demonstrate the immediate effect of the modified UCBL shoe insert. The direction and magnitude of the movement of the joints collected by the two-dimensional system suggested that the orthotic intervention had positive effects on the motions of the subtalar, ankle and knee joints. These results suggest that the use of a modified UCBL shoe insert for flexible flat foot subjects may reduce the magnitude and duration of abnormal pronation during the stance phase of gait and this may reduce the abnormally high stress on the plantar ligaments and lessen abnormal tibial external rotation. The reduction of the abnormal motion should also relieve the associated heel and knee pain caused by the pathomechanics. Particularly even after heel-off, without the use of an orthosis the subtalar joint continued to pronate; but with the orthosis earlier resupination of the subtalar joint was encouraged.

The result of this within-subject single measurement experiment only provided information about the immediate effect of the orthotic treatment. The natural history of the flexible flat foot has not been explained. Longitudinal studies with multiple measurements and use of a control group would be necessary. The arch index (Canavagh and Rodgers, 1987) used in this study to quantify the degree of flat foot is based on the cumulative measure of plantar pressure on the foot. As flat foot induces a lot of dynamic changes in the ankle-foot complex during gait, a measure of the transient pedobarographs during certain specific instants should be a more meaningful indicator of foot types. The accuracy of the force platform was 22 Newtons. Thus the force platform could not detect the difference between the shod and the orthotic conditions at less than this value. Devices such as in-sole pedobarograph system could be used to investigate the plantar surface loading distribution (Hennig *et al.*, 1994). The proposed method could also record data of multiple steps instead of the 'one step' approach of the single platform system (Chang *et al.*, 1994). The two-dimensional system has already detected some kinematic and kinetic changes due to the orthotic intervention. To investigate the relationship among the subtle kinematic changes at different joints a three-dimensional analysis system should be attempted.

A high degree of ligament laxity in the Hong Kong population has been reported (Cheng *et al.*, 1991). The normal values of the parameters that have been discussed may not be applicable for the Chinese children in Hong Kong. In parallel to this study, a project was started to develop the norm to classify foot shape by investigating the dynamic foot prints of subjects ranges from 4 to 18 years of age. Future work will be the establishment of an indicator for screening of the abnormal, flexible flat foot, and an evaluation of the efficiency and efficacy of the orthotic treatment in well defined flat foot subjects through the study of the relationships between plantar surface loading on the foot and three-dimensional joint motion of the lower limb. Since the plantar foot pad will not diminish fully until the age of 4 years, taking into account the development of the normal foot, an ideal longitudinal study of orthotic treatment involving subjects of at least 5 years of age should be undertaken. A parallel control group with no treatment should be established to investigate the long term effect of orthotic treatment for flexible flat foot.

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